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Inter-Segmental Coordination during a Unilateral 180° Jump in Elite Rugby Players: Implications for Prospective Identification of Injuries

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Abstract: Musculoskeletal injuries often occur during the execution of dynamic sporting tasks that involve rotation. The prescription of appropriate prevention strategies of musculoskeletal injury relies on assessments to identify risk, but current assessment tools focus on uniplanar movements. The purpose of this paper is to demonstrate the utility of the unilateral 180° jump as a potential assessment tool for injury risk in the lower body by (1) providing descriptive kinematics of the knee, thigh, and pelvis (2) conducting inter-segmental coordination analysis, and (3) comparing the knee kinematics between the dominant and non-dominant limb (NDL) during the loading (LOP) and landing phase (LAP). Elite rugby players completed one session, performing five 180° unilateral jumps on each limb while collecting kinematic data. Independent *t*-tests were used to compare peak angles of DL and NDL. Continuous Relative Phase (CRP) plots were constructed for thorax and pelvis in the transverse plane. At the loading phase, the non-dominant limb had greater peak knee abduction (ABD) (p = 0.01). At the landing phase, the dominant limb had greater peak knee adduction (ADD) (p = 0.05). At the landing phase, the non-dominant limb had greater peak knee ABD (p = 0.01). CRP plots indicate participants can utilize a thorax-led, pelvis-led, or synchronized rotational method. Bilateral asymmetries were observed, indicated by significant differences in the bilateral landing phase peak ADD/ABD, which is of particular interest considering all participants were healthy. Therefore, additional research is needed to determine thresholds for injury risk during rotational tasks.

Keywords: functional; movement; evaluation; assessment; screen

1. Introduction

Athletes in sports that involve jumping, pivoting, forceful change of direction, and contact are at a higher risk of lower-body injury [1–5]. Non-contact anterior cruciate ligament (ACL) tears are one of the most common lower-body injuries and have been largely attributed to the torsional forces associated with cutting and rotational tasks [4–6]. Rugby incorporates all of the aforementioned characteristics and it has been reported that lower limb joint/ligament injuries are the most common location and type of injury [7] occurring in both contact and non-contact situations [8]. Knee injuries result in players missing the most days from training, with ACL injuries accounting for the greatest proportion [9].

Clinicians and practitioners attempt to pre-emptively identify modifiable risk factors and address through strength, flexibility, and neuromuscular training. The identification of risk factors requires the utilization of a physical assessment tool with valid and reliable prognostic value for sporting tasks.

Pre-participation screenings are designed to determine potential intrinsic injury risk factors by identifying characteristics of the musculoskeletal system that may predispose an athlete to injury or identify incomplete recovery from a previous injury [10]. Traditionally, screening methods require a battery of tests, including static and dynamic protocols [10–16]. Static tests focus on measurements of joint range of motion, muscle strength, and muscle flexibility. The limitations of static tests are their applicability to dynamic situations, yielding limited meaningful information in the context of a dynamic sporting environment [17]. Previous research has demonstrated that scores obtained during static balance tests do not reflect scores obtained during dynamic balance tests, further indicating their limitations as injury predictors in a sporting context [16,18]. The recognition of these limitations led to the emergence of more functionally relevant or task-specific screening tests; where the functional aspect refers to the adoption and inclusion of assessments that more closely replicate activities of daily living and dynamic sports movement [19].

The purpose of functional and dynamic testing is to evaluate movements similar to sports actions that require the muscles to co-activate in integrated patterns to control a multi-joint movement [20]. A failure of the muscles to activate in a coordinated manner in the control of a movement often results in the development of a compensatory but predictively variable strategy to complete the task, which, after a period of time, can lead to injury and pain around the affected joint [11,17,21]. Functional testing is intended to enable the tester to identify 'weak links', suggesting an uncontrolled movement system (joint), within the series of linked joints [17]. Identification of these movement dysfunctions may serve as a predictor of injury and can be addressed by the appropriate professionals prior to the emergence of symptoms within a sporting context. Specifically, Mottram and Comerford [17] stated that the ability to identify uncontrolled movement strategies is imperative to enable risk management strategies to be developed, which can mitigate the propensity for (re)injury through increased localized strength and/or enhanced motor control.

Several mechanical dysfunctions have been associated with functional/dynamic assessments as injury predictors. Multiple studies have suggested that uncontrolled knee motion in the frontal plane when landing is a good predictor of ACL injury [22–24]. Additionally, compensatory strategies have been demonstrated in jump landings following injury [25,26]. The neuromuscular control associated with this form of dysfunctional, maladaptive movement pattern has been suggested as a potentially modifiable risk factor [27]. The utilization of unilateral tasks further enhances the assessment through the identification of limb asymmetry when compared to bilateral assessments [15,16,18]. However, research regarding cutting tasks (change of direction) suggests that large frontal plane knee excursions are necessary in order to complete the task and may not be indicative of uncontrolled motion [28–31]. Therefore, drop jumping landing tasks may not be appropriate for assessing injury risk during rotational sporting tasks. Designing an appropriate task to identify the predictors of rotary injury risk during dynamic sports-specific activity must include several components: high-velocity loading force, a certain degree of motor control complexity to complete the task, bilateral assessment, and multi-planar movement. A novel unilateral 180° jump task could potentially fulfill all the aforementioned requirements and better mimics the dynamic multi-planar nature of sports that involve jumping, pivoting, and forceful changes in direction (i.e., rugby, soccer, basketball, American football). Additionally, as athletes performing rotational movements are the target population for this assessment, using elite-level athletes, such as rugby players, will provide greater external validity of findings.

Considering the unilateral 180° jump as a potential physical assessment tool for rotational injury prediction in the lower body first requires a demonstration of its utility. Therefore, the purpose of this investigation is to analyze the unilateral 180° jump in elite-level rugby players by (1) providing descriptive kinematic of the knee, thigh, and pelvis, (2) conducting inter-segmental coordination analysis between the thorax and pelvis, and (3) comparing the knee kinematics between the dominant and non-dominant limb during the loading and landing phase. It is hypothesized that the greater rotational component will demonstrate greater frontal plane kinematics (greater peak abduction/adduction angles) of the knee during landing to compensate for the high torsional energy transfer through the kinetic chain. Similarly, greater peak frontal plane knee kinematics will be associated with a shoulder led transverse thorax–pelvis relative phase angle. More simply, we believe that greater knee adduction angles will be observed in individuals who utilize a shoulder led rotation pattern.

2. Methods and Materials

2.1. Participants

Fourteen male, elite rugby players from a National Union Academy (age: 20.0 ± 1.9 years; height: 184.5 ± 7.2 cm; mass 94.7 ± 12.0 kg) volunteered their participation. The inclusion criteria required that all participants were free of any current injury and had no history of serious lower limb injuries (injuries requiring surgery). To ascertain technique limb dominance, participants were asked which leg they preferred to kick a ball with. All participants indicated that their preferred kicking leg was their right leg. As a result, the right leg will be referred to as the dominant leg and the left as the non-dominant leg. All participants were 18 years of age or older. The study adhered to the guidelines set by the Declaration of Helsinki, ethical approval was obtained from the local University Ethics Committee (Edinburgh Napier University) prior to the investigation, and all participants provided written informed consent.

2.2. Procedures

Participants were requested to refrain from strenuous activity at least 24 h before attending the testing session. All participants were given verbal instructions indicating how to complete the jumping task. These included to initiate a unilateral stance, perform a countermovement action followed by a 180° jump as high as possible (maximal effort) landing on the same leg. Participants were asked to jump in a clockwise direction from their right leg and counterclockwise from their left leg. Participants were instructed not to use any arm movements during the jumps by placing "their hands on their hips". Following a demonstration of the jump, all participants practiced during an individualized warm-up to familiarize themselves with the protocol. Subsequent to the warm-up, each participant rested for 3 min before performing five single-leg 180° jumps either with their dominant or non-dominant leg (with a balanced pause between each jump), followed by five additional jumps with the other leg. The order of the starting jump leg was randomized to avoid learning effects.

2.3. Motion Analysis

A three-dimensional analysis was carried out using a 12-camera high speed (240 Hz) motion capture system (ProReflex, Qualisys AB., Gothenburg, Sweden). All participants wore fitted clothing and were barefoot to permit the accurate attachment of 25 retro-reflective markers (19 mm diameter) on the following anatomical landmarks (on both left and right sides): head of the first and fifth metatarsal bones, lateral and medial malleolus, posterior calcaneus, lateral and medial femoral epicondyles, anterior superior iliac spine (ASIS), posterior superior iliac spines (PSIS), acromion process, 4th lumbar vertebra, 10th thoracic vertebra, 7th cervical vertebra, sternum jugular notch, and xiphoid process. To reduce skin movement artifact error, clusters of 4 retro-reflective markers fixed to lightweight rigid plastic plates (Qualisys AB, Gothenburg, Sweden) were attached to both thighs and shanks to track the leg movement during the unilateral 180° jump, and a cluster of 4 markers placed on the skin around T7 was used to track thorax motion. Following the marker placement procedure, a static calibration file in the anatomical position was collected for each participant for the purpose of 3D model building and data generation.

2.4. Data Analysis

Marker trajectories were smoothed with a 6 Hz fourth-order low-pass Butterworth filter and kinematic data processed using Visual 3D^{TM} software (C-Motion Inc., Rockville, MD, USA). A power spectrum density calculation was conducted and 99% of the power was contained in the first 6 Hz, and, therefore, a 6 Hz cutoff frequency was chosen. All kinematic variables were quantified within the loading and landing phases of the unilateral 180° jump. The loading phase was defined as the instant the knee commenced flexion (following a stationary single-leg stance) until the instant of maximum knee flexion of the same leg. The landing phase was defined as the instant the landing foot made contact with the ground (determined from kinematic data when the metatarsal markers stopped their downward trajectory) until maximum knee flexion was attained.

Individual segment pose and segment coordinate systems (SCS) for each participant was generated with Visual 3D^{TM} from the static calibration trial using a multi-anatomical landmark optimized method [32] and utilizing the markers as follows: The pelvis was defined using the X-Y plane passing through the ASIS and PSIS, the origin of the SCS at the midpoint between the 2 ASIS markers, which allowed the segment X axis to be determined by the vector between the origin and the right ASIS marker. The SCS Z axis was determined by the axis perpendicular to the X-Y plane in the vertical direction, and, lastly, the SCS Y axis calculated as the cross product of the X and Z axes. All remaining segments were defined using the principle of creating a frontal plane using medial and lateral proximal and distal markers and then determining the SCS Z (vertical) axis as the unit vector directed from the distal segment end to the proximal segment end. The SCS Y axis was then perpendicular to the frontal plane and Z axis, and, finally, the X axis determined by the application of the right-hand rule. The origin of each SCS was located at the proximal end for each segment.

Knee joint angles were calculated according to the International Society of Biomechanics recommendations as the shank segment relative to the thigh segment resolved into the proximal segment SCS, using the Cardan sequence XYZ, where movement in the X plane denotes flexion (−)/extension (+), Y plane abduction (−)/adduction (+), and Z plane axial internal (+)/external (−) rotation [33]. Knee alignment in the Y plane was defined as zero when the long axes of the thigh and shank were aligned. Axial rotation of the thigh segment (Z plane) was calculated relative to the pelvis segment [33].

For both adduction/abduction and axial rotation, the non-dominant leg was multiplied by negative one to allow comparison between limbs. Pelvis motion was assessed in terms of flexion/extension, adduction/abduction, and axial rotation, which was computed as the angle of the pelvic segment relative to the fixed laboratory/global coordinate system (GCS) using Cardan sequence XYZ and the right-hand thumb rule [33]. Given the dynamic nature of the activity and magnitude of axial rotation involved in this activity, the sequence XYZ was selected instead of ZYX as described by Baker [34]. All variables were averaged across the five trials per leg for each participant; these means were then used to calculate group means.

The continuous relative phase (CRP) was calculated as a representative measure of inter-segmental coordination between the thorax and pelvis segmental rotations about the vertical (z) axis. To calculate the CRP between two segments, the phase-angle (Φ) from the phase-plane portraits of each segment were calculated. Phase-portraits were constructed with angular displacement (θ) on the horizontal (x) axis and the first derivative, angular velocity (ω) on the vertical (y) axis. Prior to calculating Φ, the phase-portrait values were normalized to the minimum and maximum values found in each axis using the protocol outlined by Li et al., [35], and to 101 data points. The normalization procedures minimize the influence of different segmental movement amplitudes [35] and allow comparisons of jumps with different temporal structures. The Φ was defined as the angle between the right horizontal and a line drawn to a specific data point (θ_i,ω_i) from the origin (0,0). The CRP was calculated as the difference between the thorax and pelvis segment angles at each of the 101 data points. Ensemble curves were produced for the individual CRP profiles by determining the mean CRP value at each of the 101 data points from the 5 jump trials. The variability in the CRP was displayed as the standard deviation of the 5 trials at each data point. The CRP provides a means to interpret both the coordination between the relative segments and its variability. This information can give insight into the relative stability (change in variability) of a pattern of movement over time, helping to identify which, if any, coordination patterns are important and common across multiple individuals. Measuring the relative phase between limb/segment movements (oscillations) has been regularly employed to examine the organization of a system at a synergistic level, as phase differences reflect the fundamental cooperation and competition evident within a movement system. The tendency of the synchronization of the reversal points of frequency-locked coupled oscillators is to adopt either an in-phase (0°) or anti-phase (180°) relationship with the movements initiating and/or terminating simultaneously [36]. According to Swinnen et al., [37] synchronization of the reversal points can be interpreted as intermittent loci of control, where reversal points act as anchors for the organization of the system. In contrast, asynchronous phase differences (i.e., 90°, 270°, etc.) are more difficult to maintain, requiring effort and considerable practice [38].

2.5. Statistical Analysis

All statistical analyses were performed using the Statistical Package for Social Sciences 14.0 (SPSS Inc., Chicago, IL, USA, 2004). The Kolmogorov–Smirnov test was used to determine the normality of the data distribution for each variable. Measures of central tendency and distribution of the data were reported as means and sample standard deviations. Paired sample *t*-tests were used to determine if any statistically significant differences existed between the mean values of the dominant and non-dominant legs for peak knee angle (flexion/extension and adduction/abduction), time to reach peak knee angle, axial rotation of the femur with respect to the pelvis and axial rotation of the pelvis with respect to the GCS in both the loading and landing phases following a 180° unilateral jump. Significance was accepted at $p \leq 0.05$ for all statistical tests. To measure the magnitude of the difference between the dominant and non-dominant legs relative to the variability, effect size (d) calculations were performed across all variables. Interpretation of the data was based on Cohen's (1992) guidelines, whereby effect sizes greater than 0.2 and less than 0.5 are considered small, greater than 0.5 and less than 0.8 are moderate, and greater than 0.8 are large. The intra-class correlation coefficient (ICC) was calculated to determine the reliability of each variable across the repeated trials. The standard error of measurement (SEM) was also calculated to assess the test's reliability, for example, a larger SEM indicates a lower test reliability and was calculated as follows:

$$SEm = SD * \left(\sqrt{1} - ICC\right)$$

3. Results

The results are presented with respect to the knee joint motion, thigh motion, pelvis motion, and inter-segmental coordination employed during the unilateral 180° jump. The ICC values (see Table 1) show a large range (ICC's Dominant Limb: Pelvis 0.03–0.83; Knee 0.34–0.92; Non-Dominant Limb: Pelvis 0.13–0.74; Knee 0.32–0.89) of test–retest reliability scores across the variables of interest, with a tendency for the dominant limb to demonstrate greater reliability across a greater number of variables.

3.1. Knee Joint Motion

During both the loading and landing phases, peak knee abduction was found to differ significantly between the legs, with the knee of the non-dominant leg abducting more than the dominant leg (see Table 2). Peak knee adduction between the legs differed significantly in the landing phase, with the knee of the dominant leg adducting more than the knee of the non-dominant leg. Peak knee adduction and peak knee abduction occurred within the early and late phases, respectively, during both the loading and landing phases. All other knee variables did not differ between the dominant and non-dominant leg within the loading and landing phases.

Table 1. Reliability measures.

Phase	Variable	Non-Dominant Limb		Dominant Limb	
		ICC [95% CI]	SEM	ICC [95% CI] (Dominant)	SEM
Loading	Peak Flexion (degs)	0.83 [0.68–0.93]	1.56	0.83 [0.68–0.94]	1.55
	Peak Adduction (degs)	0.89 [0.77–0.96]	0.63	0.83 [0.67–0.93]	1.28
	Peak Abduction (degs)	0.87 [0.74–0.95]	0.92	0.90 [0.80–0.96]	0.58
	Time Peak Add (%SC)	-	-	-	-
	Time Peak Abd (%SC)	-	-	-	-
Landing	Peak Extension (degs)	0.70 [0.47–0.87]	1.44	0.78 [0.59–0.91]	1.03
	Peak Flexion (degs)	0.73 [0.52–0.89]	2.35	0.77 [0.58–0.91]	1.96
	Peak Adduction (degs)	0.86 [0.73–0.95]	0.66	0.79 [0.61–0.92]	1.26
	Peak Abduction (degs)	0.87 [0.75–0.95]	0.81	0.92 [0.83–0.97]	0.44
	Time Peak Add (%SC)	-	-	-	-
	Time Peak Abd (%SC)	-	-	-	-

ICC 0.5–0.75 = moderate reliability, ICC 0.75–0.90 = good reliability, & ICC > 0.90 = excellent reliability. Larger SEM indicates less reliability.

Table 2. Knee motion for both loading and landing phases. Abd—Abduction; Add—Adduction.

Phase	Variable	Non-Dominant	Dominant	*p*-Value	Mean Diff [95% CI]	Effect Size (d)
Loading	Peak Flexion (degs)	−58.2 ± 9.2	−59.7 ± 9.1	0.67	1.5 [15.2, −12.2]	0.16
	Peak Adduction (degs)	−0.3 ± 5.7	4.6 ± 7.5	0.06	5.0 [14.2, −4.3]	0.75
	Peak Abduction (degs)	−8.2 ± 7.1	−0.8 ± 5.8	0.01 *	7.3 [17.0, −2.3]	1.13
	Time Peak Add (%SC)	22 ± 25	30 ± 24	0.39	7.9 [42.3, −26.4]	0.32
	Time Peak Abd (%SC)	77 ± 23	67 ± 29	0.33	9.9 [47.9, −27.9]	0.38
Landing	Peak Extension (degs)	−13.8 ± 4.7	−11.3 ± 4.7	0.16	2.5 [9.2, −4.1]	0.53
	Peak Flexion (degs)	−47.8 ± 8.7	−46.1 ± 8.5	0.61	1.7 [14.2, −10.9]	0.20
	Peak Adduction (degs)	2.8 ± 4.7	7.0 ± 6.0	0.05 *	4.2 [11.7, −3.3]	0.79
	Peak Abduction (degs)	−8.7 ± 6.2	−2.2 ± 5.5	0.01 *	6.5 [15.0, −2.0]	1.11
	Time Peak Add (%SC)	16 ± 18	23 ± 24	0.46	6.1 [36.9, −24.8]	0.29
	Time Peak Abd (%SC)	72 ± 20	62 ± 31	0.33	9.9 [47.3, −27.5]	0.39

Mean ± Standard Deviation. * Significant at $p \leq 0.05$. Cohen's d calculated for effect size (0.2–0.5 are considered small, 0.5–0.8 considered moderate, and >0.8 considered a large effect).

3.2. Thigh Motion

There was no difference in axial rotation of the thigh between the dominant and non-dominant leg during both the loading ($p = 0.89$, d = 0.04) and landing phases ($p = 0.77$, d = 0.14). During the loading phase thigh rotation for the dominant leg was 2.2 ± 5.3 degrees and the non-dominant leg was 2.4 ± 5.0 degrees. During the landing phase, thigh rotation for the dominant leg was 6.1 ± 4.5 degrees and the non-dominant leg was 5.6 ± 4.1 degrees. In preparation for executing and following the single-leg 180° jump, the thigh segments of the dominant and non-dominant legs internally rotated.

3.3. Pelvis Motion

During both the loading and landing phases, pelvis motion was not significantly different between dominant and non-dominant legs (Table 3). In preparation for taking-off, the pelvis was found to extend (sagittal plane), adduct (frontal plane), and externally rotate (transverse plane). On landing, the pelvis was found to extend, abduct, and internally rotate.

Table 3. Pelvis alignment for both loading and landing phases. Flex—Flexion; Ext—Extension; Add—Adduction; Abd—Abduction; Rot—Rotation.

Phase	Variable (degs)	Left	Right	*p* Value	Mean Diff [95% CI]	Effect Size (d)
Loading	Flex/Ext	−12.0 ± 7.5	−9.2 ± 5.5	0.27	2.8 [12.1, −9.3]	0.43
	Add/Abd	−3.6 ± 3.1	−4.3 ± 2.4	0.46	0.7 [4.3, −3.6]	0.26
	Axial Rot.	4.4 ± 6.0	0.8 ± 4.6	0.09	3.6 [11.1, −7.5]	0.68
Landing	Flex/Ext	−1.2 ± 3.8	−1.1 ± 3.1	0.95	0.1 [6.5, −6.4]	0.03
	Add/Abd	−7.8 ± 4.1	−7.0 ± 3.3	0.60	0.8 [6.7, −5.9]	0.22
	Axial Rot.	−18.0 ± 9.0	−18.7 ± 8.0	0.82	0.7 [12.9, −12.2]	0.08

Mean ± Standard Deviation. Cohen's d calculated for effect size (0.2–0.5 are considered small, 0.5–0.8 considered moderate and >0.8 considered a large effect).

3.4. Inter-Segmental Coordination

The three exemplar CRP plots presented (Figure 1) denote differences in pelvis/thorax segmental coordination for both the dominant and non-dominant take-off legs within the 180° jump. As shown in Figure 1, all participants tended to remain in-phase with minimal variability during the loading/take-off phase of the jump irrespective of the take-off leg. All three participants remain in-phase with low variability throughout the unilateral 180° jump when taking off from the non-dominant leg. Participant C employs a similar pattern of inter-segmental coordination (and variability) irrespective of the take-off leg. Participants A and B show a tendency to move out-of-phase and demonstrate high levels of CRP variability throughout the landing phase of the unilateral 180° jump.

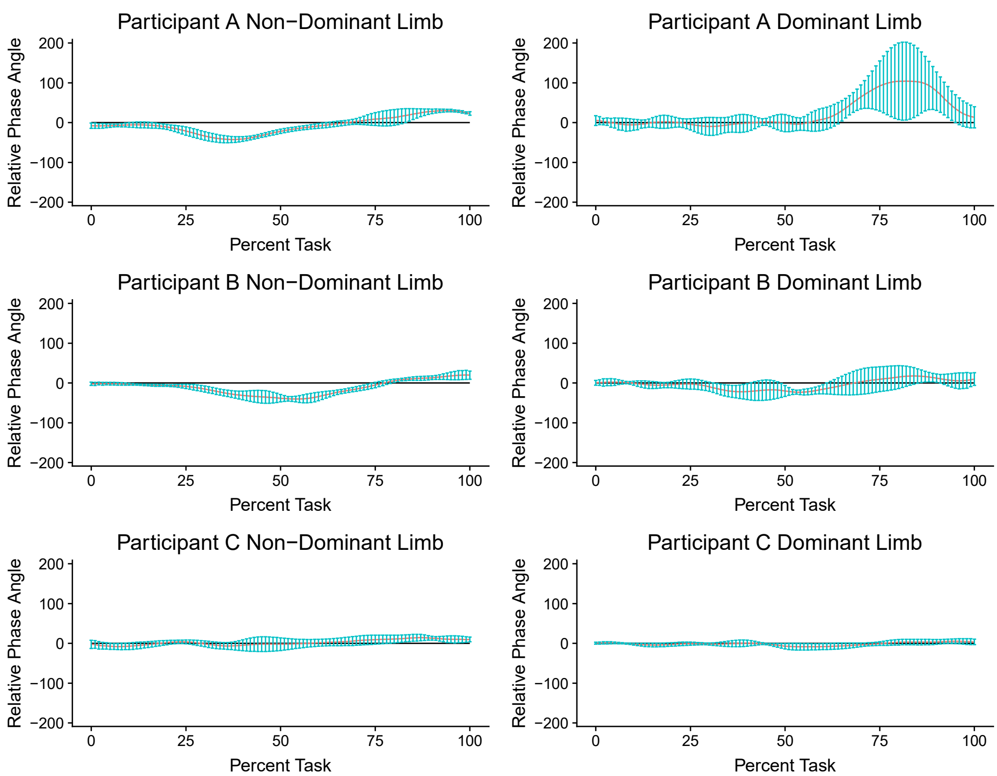

Figure 1. Exemplar continuous relative phase plots.

The solid red line represents the mean relative phase angle of all five jumps with a phase angle of 0 representing completely in phase or both the thorax and pelvis rotating together at the same rate. A phase angle > 0 (positive angle) indicates that the thorax is rotating ahead or leading the pelvis. A phase angle < 0 (negative angle) indicates the pelvis is rotating ahead or leading the thorax. The loading phase is represented by the first ~20%. The landing phase is represented by the last ~50%.

Vertical error bars indicate the standard deviation at points of the CRP plot. Therefore, larger vertical bars indicate greater movement variability jump to jump, while smaller bars represent less variability.

In phase is represented by a relative phase angle of 0. Out of phase (thorax-led) is represented by a positive relative phase angle.

4. Discussion

The principal aim of this study was to demonstrate the utility of the unilateral 180° jump as a potential physical assessment tool by providing descriptive kinematic data. All participants demonstrated similar knee and thigh motion and pelvis motion. During the loading phase there were no significant differences between limbs, with all participants executing internal rotation of the thigh while extending, adducting, and externally rotating the pelvis. These movements portray a "counter movement" pattern opposite to the desired action of rotation, by loading the system up similar to the downward movement (knee flexion) preceding forceful knee extension one would observe in a vertical jump. This rotational "counter movement" pattern may be performed in an attempt to load the associated musculature and utilize strain energy to facilitate the stretch-shortening cycle. Previous research has indicated hip flexion with hip internal rotation has a greater correlation with large knee valgus moments [29], therefore the observed external rotation of the hips in healthy athletes may be a subconscious attempt to reduce potentially deleterious forces at the knee.

During the landing phase there were no significant differences between limbs for thigh rotation ($p = 0.77$, d = 0.14) or pelvis movement (Flex/Ext [$p = 0.95$, d = 0.03], Add/Abd [$p = 0.6$, d = 0.22], Axial Rotation [$p = 0.82$, d = 0.08]). For all participants, the thigh internally rotated and the pelvis extended, abducted and internally rotated. For cutting tasks with up to a 110° change in direction, hip internal rotation in conjunction with hip abduction was associated with larger knee valgus moments [30,31]. The larger the degree of rotation the greater the rotational forces [30], indicating a necessity to execute the task and not necessarily a greater risk of injury. Since the participants of this study were healthy elite-level athletes, the pelvis moving in line with the rotation of the jump may have been a strategy to reduce torsional forces experienced at the knee and ankle by dissipating energy at a link in the kinetic chain with a greater range/number of degrees of freedom (DOF).

There was a significant difference in the frontal plane knee motion during the landing phase between limbs (Add [$p = 0.05$, d = 0.79], Abd [$p = 0.01$, d = 1.11]). Greater knee adduction was observed in the dominant limb (~4.2° greater), whereas greater knee abduction was observed in the non-dominant limb (~6.5° greater). Moreover, it should be noted that these specific findings had some of the strongest ICC values and lowest SEM values (see Table 1). This finding indicates that a bilateral asymmetry is observable even in healthy athletic populations. Due to the cross-sectional nature of this study, it is impossible to indicate which strategy is more beneficial. Uncontrolled knee movement during landing has been associated with greater ACL injury risk in uniplanar drop jump tasks [23] but rotational tasks require greater knee frontal plane excursions to complete [29–31]. Therefore, prospective evidence is needed to indicate which motion contributes to a greater risk of injury. Furthermore, one motion is most likely not good or bad, but stresses different soft tissue structures more greatly (for example, abduction: ACL, adduction: MCL). Likewise, how the system (body) moves as a whole is more important for understanding the potential injury risk.

Visual qualitative assessment of CRP plots indicated that there was little variation between participants during the loading/take-off phase (refer to the first 20% in Figure 1). Moreover, the relative phase angle between the shoulders and the pelvis for the majority of the participants was very close to 0°. A phase angle at or close to 0° indicates that the shoulders and pelvis were in-phase or moving together. Coordinated coupling of the upper and lower body segments during the loading phase may be a result of the population tested. Healthy athletes presumably have better motor control and coordinate their torsos in a synergistic fashion with their lower extremities to improve mechanical efficiency by reducing counter-productive movement. Furthermore, this may demonstrate torso control by the participants to reduce torsional forces experienced at the knee that has been observed in cutting tasks when torso lean/excursion is excessive [28]. Likewise, the amelioration of the torsional force and modulation of body segments may help maintain the center of gravity within the base of support so that the individual successfully executes the task (does not fall over). Prospective research should be conducted on unhealthy or non-athletic populations to determine if uncontrolled torso movement can be observed during the loading phase of a unilateral 180° jump task. Investigating 'less athletic' or impaired populations will further elucidate motor control capabilities, differentiating key differences in task execution compared to elite-level athletes when performing this specific task.

Whereas the loading/take-off phase exhibited relative uniformity between participants, the greatest amount of variation was observed during the landing phase of the 180° jump (refer to last 50% of Figure 1). CRP plots displayed three common landing strategies amongst participants: thorax (shoulder)-led movement, pelvis-led, or simultaneous in-phase landing. Although these three strategies were performed, it should be noted that each individual only utilized one of them and with varying degrees of phase coupling. In some cases, individuals were close to a relative phase angle of 180°, which represents an anti-phase movement (similar to arms swinging in separate directions at the same rate during normal gait). Furthermore, the phase coupling between the thorax and pelvis was not constant, meaning the rates at which the two segments rotated in conjunction with one another were at different rates throughout the second half of the unilateral 180° jump task, suggesting a dissociation between the segments during the landing portion of the task.

CRP best describes the passage of energy through a system [39,40], in this case, the human body. Each participant utilized a slightly different strategy to dissipate the high torsional energy in order to execute the task and land successfully. The fact that many strategies with different degrees of variation were observed demonstrates the principle of equifinality of movement solutions due to the vast number of DOF [41,42]. Although there are multitudes of strategies to complete the task, it stands to reason a shoulder-led strategy may be the most deleterious for structures around the knee. If the shoulders lead with the pelvis following, upon landing when a high load rate force is transmitted up the kinetic chain the shoulders will have little range left for further excursion to enable energy dissipation. Instead, it could likely result in a resultant force transmitted back down to attempt to be dissipated by the pelvis movement. This could result in greater hip flexion and hip internal rotation, which has been linked to greater knee valgus moments [29–31]. A pelvis-led strategy may enable better dissipation of excessive energy because it can pass up through the entire kinetic chain. However, because this study is cross-sectional this theory is conjecture on the part of the authors. Similar to research involving drop jumps [23,43,44], future research could yield information regarding which strategy portends greater injury risk and what relative phase angles indicate problematic upper torso and lower body coupling during rotational sporting tasks.

Further evidence to support this approach can be observed in the test–retest reliability (ICC) scores (see Table 1). Large variability can be observed across the ICC scores, and can be interpreted as being indicative of the presence of variability in specific movements, thus, enabling the production of more constrained, and reliable movements in key structures (depending on the movement strategy employed). There are some limitations to this study, mainly, the small sample size (n = 14). However, the use of elite-level athletes (rugby players) not only strengthens the external validity of this investigation (the application for athletes), the smaller sample size is representative of the smaller percentage of the population that achieved elite athlete level status (professional). Furthermore, for significant findings the effect sizes were moderate to large (>0.79) (see Table 2) and the reliability measures (ICC [0.79–0.92] and SEM [0.44–1.26]) for those that were significant were moderate to strong (see Table 1). Considering the knee is a major focal point of injury risk assessment, these aforementioned values support the unilateral 180° jump as a potential rotary injury risk assessment tool.

5. Practical Application

The unilateral 180° jump provides a plethora of biomechanical data. Firstly, the unilateral 180° jump task can be performed in a confined space making it a more feasible alternative to performing cutting tasks when assessing lower body rotary risk. Requiring the participant to jump as high as possible could provide kinetic data and a simultaneous metric on vertical jump performance. Although jump heights will not be as large in magnitude as a traditional countermovement jump, frequently, jumps are made off a single leg and unbalanced (not having time to properly align and set the body) during play, thus, the 180° jump task replicates scenarios experienced in sport. Furthermore, the requirement of maximal jump height forces the participant to utilize muscle contraction velocities that are similar to game conditions, unlike other previously mentioned assessments [2,11,12,18,45]. Likewise, landing from a maximal vertical height while rotating will exert a multi-planar loading force (vertical and torsional) typical to athletes in sports such as soccer, rugby, football, or basketball. Moreover, landing unilaterally while rotating more closely mimics one of the most common mechanisms of ACL injury as compared to landing from a uniplanar jump task [4].

Future studies should investigate the unilateral 180° jump with pathological and non-pathological populations prospectively (over the course of an athletic season) to associate movement patterns with injury predictors. Due to the multiple variables that can be collected at once with the unilateral 180° jump and the ability to visually assess the movement, the task has the potential to be a valid and efficient objective prognostic physical assessment tool that may provide greater sensitivity for identifying rotational injury risk.

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